# Simulation study of a real-world far-side crash

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## I. INTRODUCTION

Previous research on side-impact vehicle collisions has shown that they will continue to be relevant for occupant protection in terms of fatalities and serious injuries [1]. Thus, further research on side-impact collisions is of high relevance [2-4]. Since 2020, the evaluation of side-impact protection has been included in the rating of new cars by Euro NCAP [5]. However, the current assessment of side-impact protection still relies on Worldwide Harmonized Side Impact Dummies (WorldSID), which have shown limitations in both biofidelity and prediction of thoracic or abdominal injury compared with post-mortem human subject (PMHS) tests [6-9]. These limitations and the need to consider anthropometric diversity could be addressed by the use of human body models (HBMs). The objective of this research was to identify parameters with significant influence on the kinematics and injury patterns for the reconstruction of a real-world crash. The very first results of this study are shown in this paper.

## II. METHODS

The simulation was based on a collision of a 2013 Volkswagen (VW) Polo V and a 2005 Peugeot 307 SW. The front of the Peugeot collided with the passenger side of the Polo at an angle of approximately 80–90 degrees. The collision was reconstructed with PC-Crash [10]. The far-side occupant was a 58-year-old female, who sustained a lateral condyle fracture of the right tibia and a minor injury to the head. The occupant was simulated with the VIVA + model 50F-seated [11]. The vehicle interior was based on the generic vehicle interior model version 2.0 developed by Iraeus *et al*. [12]. For the far-side load case, relevant vehicle structures, such as the centre console and generic foam side structure on the seat, were adapted to the vehicle interior geometry of the VW Polo. A simulation study with parameter variation was performed [**dV**: Vehicle change in velocity (10–30 km/h); **DUR**: Pulse duration (60–140 ms); **PDOF_H**: Principal direction of force - horizontal pulse angle (70–95 °); **PTForce**: Seatbelt pretension force (0.5–2.0 kN); **KNEEFRC**: Linear stiffness of knee impact area (2.0–7.0 kN/100 mm def)], resulting in a total of 30 different crash configurations. Simulations were performed with LS-Dyna R12. The generated data were analysed with Dynasaur and a Python script [13]. Body regions relevant for far-side occupants were examined [14].

## III. INITIAL FINDINGS

Parameter configurations with different PTForce values had no effect on the injury risk in the body regions assessed. Considering the further parameter variations, the highest HIC15 was measured at a dV=30 km/h. Also for the head trajectories, dV=30 km/h was causing the biggest impact with the largest head deflection in positive y-direction throughout the accident simulations. While dV=25 km/h represented a similar trajectory, the head experienced a smaller deflection mainly in positive y-direction. The highest fracture probability in the costal arch was observed at dV=30 km/h as well, resulting in the highest maximum principal strain (MPS) in eight ribs bilateral (R1-2, R4-5, R8-10, R12), two right ribs (R3, R6), and one left rib (R11). At 34%, right R1 was at highest risk of injury at dV=30 km/h, twice as much as left R1, with an injury probability of 12.6%. Between dV 20 km/h and 25 km/h, the MPS more than quintupled in right R1, while it only doubled in left R1. For the Tibia, the injury risk doubled from dV 10–12 km/h and more than tripled from dV=20–25 km/h. The MPS of the distal tibiae, tibia shafts and proximal tibial ends were highest for dV=30 km/h. The variation of DUR resulted in mostly congruent head trajectories, except for the head deflection length in the positive y-direction. As DUR decreased, the head excursion in positive y-direction increased. Thus, the head was deflected about 50 mm more at DUR=60 ms than at DUR=80 ms. With decreasing DUR, MPS and rib injury probability increased, significantly more on the right body side than on the left. The injury probability of the tibia increased with longer DUR, peaked at DUR=100 ms and then decreased steadily with increasing DUR. Variation of PDOF_H impacted the head trajectories but did not result in any change of HIC15 values. PDOF_H of 70° showed the largest head excursion in negative x-direction, while the angle change to 95° was seen to result in the largest deflection of the head in the positive x-direction. Right R1-3 had the highest injury risk at PDOF_H 85°. The MPS of ribs 4-9 on both sides decreased over an angle

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change of 70–90°. For left R10-12, MPS steadily decreased over PDOF_H 70–95°, whereas for right R10-12, MPS increased over larger PDOF_H angles and peaked at PDOF_H 90–95°. The Tibia injury risk had its highest value at PDOF_H 90°. By increasing PDOF_H angles, the risk of injury initially dropped, increased with PDOF_H 85°, and dropped again slightly after PDOF_H 90°. KNEEFRC had no effect on the rib cage injury risk, but variation showed a significant effect on HIC15 and a small effect on proximal tibia MPS. In accordance with the real-world accident, very low injury metrics were observed for the investigated body region.

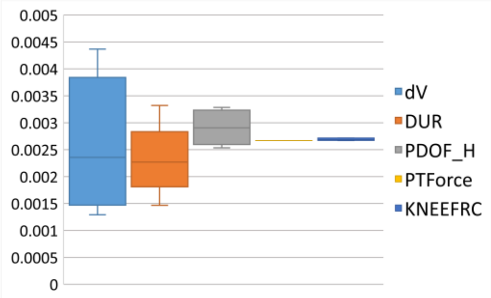


Fig. 1. Maximum principal strain of right tibia.

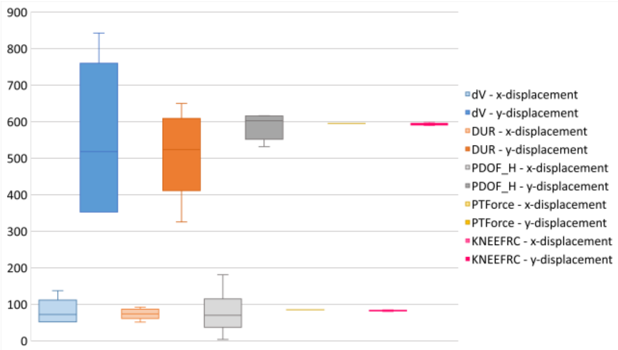


Fig. 2. Head excursion (mm).

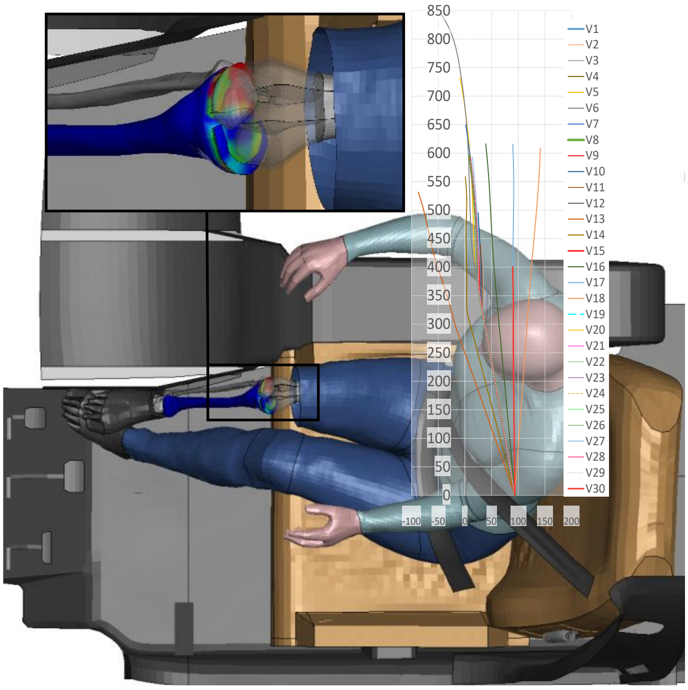

Fig. 3. Head trajectories and right tibia fringe plot (MPS).

## IV. DISCUSSION

Based on the parameter simulations and their results, an influence of PTForce on the risk of injury in the present far-side scenario could be ruled out. Even small changes of dV had a great influence on the injury probability of the examined body regions. This observation was also applicable to DUR, which is why these two parameters seem to be of high relevance for the correct accident reconstruction and simulation of real-world accidents. As KNEEFRC had an influence on HIC15 and tibial injury probability, it is also important to derive KNEEFRC as accurately as possible from the real accident. Overall, it can be stated that the simulated injuries corresponded to the injuries of the real accident. However, the injury risk also depends on parameter settings that are difficult to derive from real accidents. The reason for this can be incomplete databases or lack of information from vehicle manufacturers. Since some details, such as the seating posture or anthropometry of the occupant in question were not available, an average was assumed. These limitations give reason for future research to quantify these uncertainties. It is planned to review further real-world accidents in detail, to reproduce them with crash tests and to perform more extensive parameter studies in order to identify important factors, their interaction and impact on the injury risk of far-side occupants. As the risk curve for tibial injuries encompasses the entire bone, but in the present case, a closer examination of the tibial head would have been ideal, there is also an apparent need to develop a new risk curve for the proximal end of the tibia.

## V. REFERENCES


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